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(54) Title: PHASE-CONTRAST MRI USING PHASED-ARRAY MULTICOIL

(57) Abstract

MRI signals detected by a phased array of multicoils (1, 2,..Nc) are combined (20) to provide phase contrast image signals with improved signal-to-noise ratio. Detected complex image signals are obtained for different magnetic gradients and are combined to obtain phase, magnitude, and magnitude weighted image signals for imaging.

PHASE-CONTRAST MRI USING PHASED-ARRAY MULTICOIL

BACKGROUND OF THE INVENTION

        This invention relates generally to phase contrast
magnetic resonance imaging (MRI), and more particularly the
invention relates to phase contrast MRI using a multicoil
phased array for magnetic resonance signal detection.

        As described by Pelc et al. ("Phase Contrast Cine
Magnetic Resonance Imaging" Magnetic Resonance Quarterly
Vol. 7 No. 4, pp. 229-254, 1991), phase contrast MRI refers to
a family of MR imaging methods that exploit the fact that
spins that move through magnetic field gradients obtain a
different phase than static spins, enabling the production of
images with controlled sensitivity to flow. Among the
advantages of phase contrast MRI are its adjustable
sensitivity to flow and, in some cases, the quantitative
nature of the resulting data. In the most common method, two
data sets are acquired with a different gradient first moment
in one direction, and the pixel by pixel phase difference $\Delta\phi$
in the resulting images is used to measure the velocity, V, in
the direction of the gradient change: $\Delta\phi = \gamma \Delta M_1 V$, where ΔM_1 is
the change in gradient first moment and γ is the gyromagnetic
ratio. Four measurements can be used to measure all
components of velocity. Sometimes, for aesthetic reasons
related to the very high noise in $\Delta\phi$ in regions of low signal,
"magnitude weighted" or "magnitude masked" velocity images
with pixel intensities proportional to $M\Delta\phi$ are produced, where
M is the magnitude image.

        Phased array multicoils can be used in MRI to
improve the signal-to-noise ratio (SNR) of images. See, for
example, *Roemer et al.* "The NMR Phased Array", Magnetic
Resonance in Medicine, Vol. 16, pp. 192-225, 1990; *Hayes et
al.* "Noise Correlations in Data Simultaneously Acquired from
Multiple Surface Coil Arrays" Magnetic Resonance in Medicine,
Vol. 16, pp. 181-191, 1990; and *Hayes et al.* "Volume Imaging

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with MR Phased Arrays" Magnetic Resonance in Medicine,
Vol. 18, pp. 309-319, 1991. Basically, the multicoils allow
large field of view imaging with the SNR of a small surface
coil. Separate images are acquired and reconstructed from
5   each of the elements in the multicoil array. These separate
images are then combined into a single image, with each coil
dominating the spatial regions where its SNR is the highest.
Although various algorithms for this combination process have
been proposed, the most commonly used methods use the
10   magnitude of the pixel values in the individual images. For
example, one very common method forms images as the square
root of the sum of squares of the individual images.

        The combination of phase contrast techniques and
phased array coils would be very desirable in order to improve
15   the SNR of the velocity-sensitive images. However, a simple
and efficient method for combining the information received
from the various coils that retains a proportionality with
velocity is needed.

20   SUMMARY OF THE INVENTION
        In accordance with the invention, the complex image
data (Z_i) from each coil (i) in a multicoil phased array
including the magnitude (M_i) and the phase (ϕ_i) are employed
along with specific coil gain factors (W_i) to generate phase,
25   magnitude, and magnitude-weighted images.
        More particularly, let Z_{1i} and Z_{2i} be the complex
images reconstructed from coil i with sequences 1 and 2, and
magnitude image, M_i, equal

$$M_i = (|Z_{1i}| + |Z_{2i}|)/2$$

30   and phase image, ϕ_i, equal

$$\phi_i = phase(Z_{1i}Z_{2i}^*)$$

where * indicates the complex conjugate and
            phase (Z) = arctan2(imag(Z),real(Z)).
Further, let W_i be the gain factor for coil i.

Then, in accordance with the invention, the phase, magnitude, and magnitude-weighted image are given as:

$$M = \sqrt{|C|}$$

$$\Delta\phi = \text{arctan2 (imag(C), real (C))}$$

mag weighted image $= M \, \Delta\phi$

5   where

$$C = \sum \left[(Z_{1i} \, Z_{2i}^{*})/W_i^2 \right]$$

The invention presents a simpler and more optimal method in using phase contrast MRI with a phased array multicoil than heretofore known.

10   The invention and objects and features thereof will be more readily apparent from the following detailed description and appended claims when taken with the drawing.

BRIEF DESCRIPTION OF THE DRAWINGS

15   Figs. 1A-1D illustrate the arrangement of MRI apparatus and magnetic fields generated therein.

Fig. 2 is a function block diagram of MR imaging apparatus.

Fig. 3 is a schematic view of an MRI phased array as

20   used in NMR apparatus.

DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring now to the drawings, Fig. 1A is a perspective view, partially in section, illustrating coil

25   apparatus in MR imaging system and Figs. 1B-1D illustrate field gradients which can be produced in the apparatus of Fig. 1A. Briefly, the uniform static field B_0 is generated by the magnet comprising the coil pair 10. A gradient field G(x) is generated by a complex gradient coil set which can be

30   wound on the cylinder 12. An RF field B_1 is generated by a saddle coil 14. A patient undergoing imaging would be positioned along the Z axis within the saddle coil. In Fig. 1B an X gradient field is shown which is parallel to the static field B_0 and varies linearly with distance along the X

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axis but does not vary with distance along the Y or Z axes.
Figs. 1C and 1D are similar representations of the Y gradient
and Z gradients fields, respectively.

5   Fig. 2 is a functional block diagram of conventional
imaging apparatus. A computer 20 was programmed to control
the operation of the MRI apparatus and process FID signals
detected therefrom. The gradient field is energized by a
gradient amplifier 22 and the RF coils for impressing an RF
magnetic moment at the Larmor frequency is controlled by the
10   transmitter 24 and the RF coils 26. After the selected nuclei
have been flipped, the RF coils 26 are employed to detect the
FID signal which is passed through the receiver 28 and then
through digitizer 30 for processing by computer 20.

           Fig. 3 is a schematic representation of a
15   conventional phased array for use in MRI, as disclosed in
Roemer U.S. Patent No. 5,086,275. The NMR phased array 10
includes a plurality of radio frequency (RF) receiver coils 12
(coils 1 through N_c) defining an imaging volume for the NMR
imaging of a sample, such as for the NMR medical diagnostic
20   imaging of a human spine. The separate surface coils 12 are
identically configured and are arranged in closely spaced
relationship with overlapping field of view (FOV), but with
substantially no interaction between adjacent coils. The
coils 12 are adapted as part of NMR imaging apparatus such as
25   shown in Figs. 1 and 2, to simultaneously receive a different
one of a plurality of NMR response signals, each evoked from
an associated portion of the sample enclosed in the imaging
volume. As shown, each coil 12 has its own processing channel
14 including receiver circuitry 15 and an analog to digital
30   converter 16. This data processing setup provides for
constructing a different NMR image for each channel 14 of a
sample portion from the NMR response signals received by the
associated coil 12 for that channel 14, and for subsequently
combining the plurality of different images thus constructed,
35   on a point by point basis, in the image domain, to produce a
single, final NMR image of all sample portions from which an
NMR signal was received by any of the coils 12. As further
described by Roemer, the optimal combination or weighting of

signals from the individual coils $1-N_C$ in the array 10 to
achieve high signal-to-noise ratio (SNR) is dependent on the
location of a particular volume element (voxel). This is
because the signal of each RF receiving coil C_i is sensitive
5   to nuclear spins in proportion to the field B_i created by the
coil, whereas the noise is "white noise" uniformly distributed
over the image. Hence, the resultant SNR is a function of
position.

        As noted above, phase contrast magnetic resonance
10   imaging (MRI) uses flow dependent contrast to produce images.
In the most common method, two data sets are acquired with a
different gradient first moment in one direction and the pixel
by pixel phase difference $\Delta\phi$ in the resulting images is used
to measure the velocity V in the direction of the gradient
15   change: $\Delta\phi = \gamma\Delta M_1 V$, where ΔM_1 is a changing gradient first
moment and γ is the magnetic ratio.

        In accordance with the invention, a simple and
efficient method is provided for combining the information
received from the multicoils that retains the proportionality
20   with velocity.

        Consider now phase contrast imaging of a single
velocity component (two measurements) and let Z_{1i} and Z_{2i} be
the complex images reconstructed from coil i with sequences 1
and 2. Similarly, let

$$M_i = (|Z_{1i}| + |Z_{2i}|)/2$$

$$\phi_i = phase(Z_{1i} \, Z_{2i}{}^*)$$

25

be the magnitude and phase images from coil i, where *
indicates the complex conjugate and
                phase(Z) = arctan2(imag(Z),real(Z)).
The set $\{\phi_i\}$ can be thought of as estimates of the true
30   velocity induced phase shift $\Delta\phi$. As long as the coils are
uncoupled the noise in $\{\phi_i\}$ are uncorrelated but have

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different variances. The variance in ϕ_i, σ^2_i, is inversely proportional to M_i^2:

$$\sigma^2_i \propto \frac{1}{M_i^2 / W_i^2}$$

5   The linear combination of $\{\phi_i\}$ which minimizes the variance uses a weight for coil i which is inversely proportional to σ^2_i. Thus, one improved algorithm for estimating $\Delta\phi$ is:

$$\Delta\phi = \frac{\sum M_i^2 \phi_i / W_i^2}{\sum M_i^2 / W_i^2}$$

A magnitude weighted image is formed as $M\Delta\phi$ with:

$$M = \sqrt{\sum M_i^2 / W_i^2}$$

10   Note that the last equation is the conventional magnitude multicoil image. The magnitude weighted phase image is:

$$M\Delta\phi = \frac{\sum M_i^2 \phi_i / W_i^2}{\sqrt{\sum M_i^2 / W_i^2}}$$

This method requires calculating a separate magnitude and phase shift (arc tangent) image for each coil.

15   An alternate and preferred method combines the data from the multiple coils prior to the arc tangent. Each complex image, e.g., Z_{1i}, contains phase shifts from sources other than the desired velocity component. These undesired phase shifts are also contained in Z_{2i}, so that the phase in

20   Z_{1i}/Z_{2i} is, ideally, due only to the desired motion. The phase of $(Z_{1i} Z_{2i}^*)$ is exactly the same. For this reason, as well as computational efficiency and automatic mapping into a 2π range, the above equation for ϕ_i is often used to compute the phase shift. The quantity $(Z_{1i} Z_{2i}^*)$ is a unit vector in

25   the direction ϕ_i multiplied by (approximately) M_i^2. Each coil provides a similar vector, and, except for noise, all point in

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the same direction: $\Delta\phi$. Thus, $(Z_{i1} Z_{2i}^*)$ can be viewed as a scaled (weighted) estimate of $e^{i\Delta\phi}$. One might therefore ask: what is the ideal linear combination of multiple estimates of $e^{i\Delta\phi}$. To first order, the estimates should have weights proportional to M_i^2/W_i^2. Thus we form:

$$C = \sum [(Z_{1i} Z_{2i}^*) / W_i^2]$$

$$M = \sqrt{|C|}$$

$$\Delta\phi = \text{arctan (imag(C), real(C))} \qquad \text{(11b)}$$

$$\text{mag weighted image} = M \Delta\phi \qquad \text{(11c)}$$

There have been described methods and apparatus for phase contrast MRI with a phased array multicoil in which the information from the various coils can be combined while retaining the proportionality with velocity. While the invention has been described with reference to specific embodiments, the description is illustrative of the invention and is not to be construed as limiting the invention. Various modifications and applications may occur to those skilled in the art without departing from the true spirit and scope of the invention as defined by the appended claims.

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WHAT IS CLAIMED IS:

1.   In a multicoil magnetic resonance imaging
system, a method combining signals detected by multicoils for
phase-contrast imaging comprising the steps of
a)   detecting complex image signals Z_{1i} and Z_{2i}
with each coil, i, in two excitation sequences with different
magnetic gradients,
b)   obtaining a combined image C, as the summation
of $(Z_{1i}Z_{2i}^{*})/W_{i}^{2}$ where
Z_{2i}^{*} is the complex conjugate of Z_{2i}
W_{i} is a gain factor for coil, i; and
c)   forming an output image dependent on C.

2.   The method as defined by claim 1 and further
including the step:
d)   obtaining phase, ϕ, magnitude, M, and magnitude
weighted image, $M\Delta\phi$, as follows:

$$M = \sqrt{|C|}$$

$$\Delta\phi = \text{arctan2 } [\text{imag } (C), \text{real}(C)]$$

3.   The method as defined by claim 1 and further
including the step of imaging velocity in a region using said
combined image.

4.   The method as defined by claim 3 wherein said
magnetic gradients vary in one direction and said velocity is
in the direction of gradient change.

5.   The method as defined by claim 1 and further
including the steps of detecting two additional complex image
signals, Z_{3i} and Z_{4i}, and said magnetic gradients vary in
three dimensions, said complex image signals providing
components of velocity in said three dimensions.

6.   In apparatus for phase-contrast magnetic
resonance imaging, the combination comprising:

a)   means for generating a magnetic field,

b)   means for generating at least one magnetic
gradient field,

c)   multicoils for detecting complex image signals
Z_{1i} and Z_{2i} with each coil, i, in two excitation sequences
with different magnetic gradients; and

d)   means for obtaining a combined image C, as the
summation of $(Z_{1i}Z_{2i}{}^{*})/W_i{}^2$ where

$Z_{2i}{}^{*}$ is the complex conjugate of Z_{2i}

W_i is a gain factor for coil, i; and

e)   means for forming an output image dependent on
C.

7.   The combination as defined by claim 6 and
further including

f)   means for obtaining phase, ϕ, magnitude, M, and
magnitude weighted image, $M\Delta\phi$, as follows:

$$M = \sqrt{|C|}$$

$$\Delta\phi = \mathrm{arctan2}\ [\mathrm{imag}\ (C), \mathrm{real}(C)]$$

8.   The combination as defined by claim 6 and
further including means for imaging velocity in a region using
said combined image.

9.   The combination as defined by claim 6 wherein
said magnetic gradients vary in one direction and said
velocity is in the direction of gradient change.

10.   The combination as defined by claim 6 wherein
said multicoils detect two additional complex image signals,
Z_{3i} and Z_{4i}, and said magnetic gradients vary in three
dimensions, said complex image signals providing components of
velocity in said three dimensions.

11.   In a multicoil magnetic resonance imaging
system, a method of combining signals detected by multicoils
for phase-contrast imaging comprising the steps of

a)   detecting complex image signals Z_{1i} and Z_{2i}
with each coil, i, in two excitation sequences with different
magnetic gradients,

b)   determining velocity induced phase shift image,
$\Delta\phi$, as follows:

$$\Delta\phi = \frac{\sum M_i^2 \phi_i / W_i^2}{\sum M_i^2 / W_i^2}$$

where $M_i = (|Z_{1i}| + |Z_{2i}|)/2$

$\phi_i = \text{phase } (Z_{1i} \, Z_{2i}{}^{*})$

$* = \text{complex conjugate}$

$W_i = \text{coil gain factor}$

12.   The method as defined by claim 11 and further
determining magnitude weighted phase image as follows:

$$M\Delta\phi = \frac{\sum M_i^2 \phi_i / W_i^2}{\sqrt{\sum M_i^2 / W_i^2}}$$

13.   The method as defined by claim 12 and further
including the step of imaging velocity in a region using said
magnitude weighted phase image.

14.   The method as defined by claim 11 and further
including the step of imaging velocity in a region using said
magnitude multicoil image.

15.   The method as defined by claim 11 wherein said
magnetic gradients vary in one direction and said velocity is
in the direction of gradient change.

16.   The method as defined by claim 11 and further
including the steps of detecting two additional complex image
signals, Z_{3i} and Z_{4i}, and said magnetic gradients vary in

three dimensions, said complex image signals providing
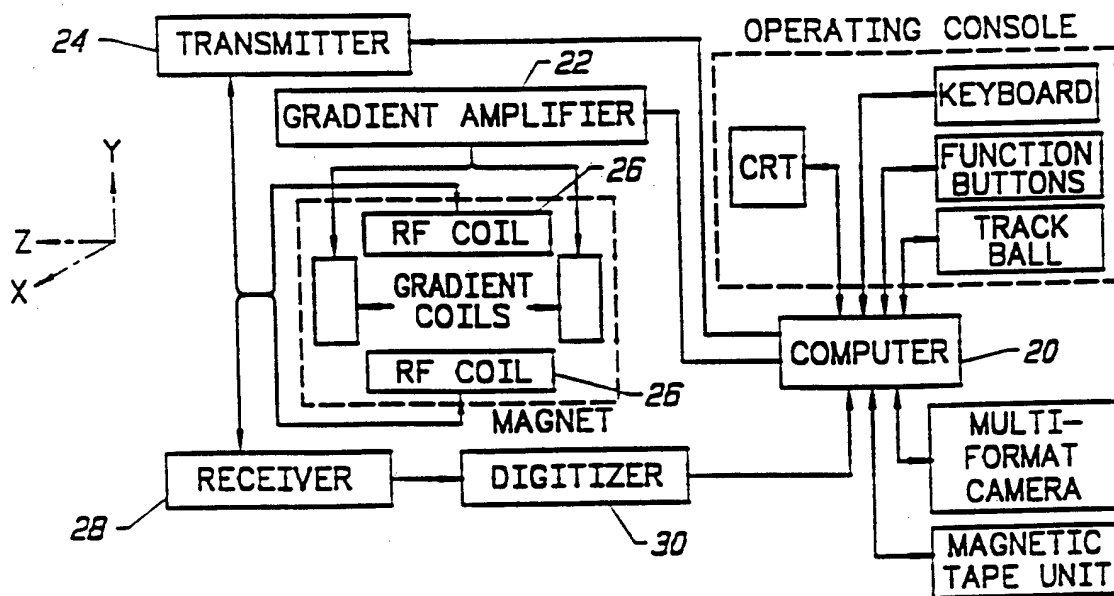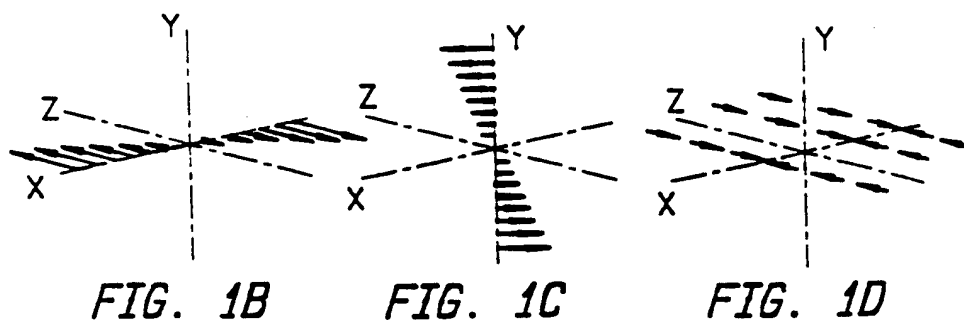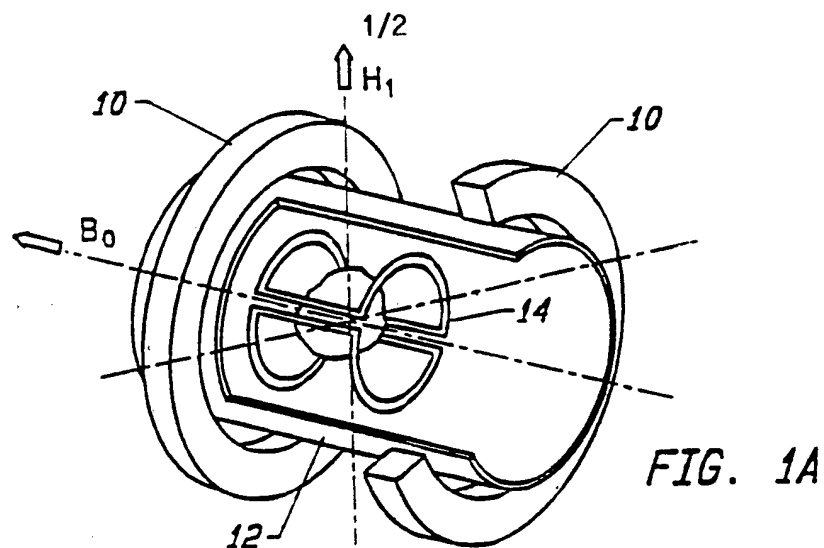components of velocity in said three dimensions.

        17.   In apparatus for magnetic resonance imaging,
the combination comprising:
            a)   means for generating a magnetic field,
            b)   means for generating at least one magnetic
gradient field,
            c)   multicoils for detecting complex image signals
Z_{1i} and Z_{2i} with each coil, i, in two excitation sequences
with different magnetic gradients, and
            d) means for determining velocity induced phase
shift, $\Delta\phi$, and M, magnitude multicoil image, as follows:

$$\Delta\phi = \frac{\sum M_i^2 \phi_i / W_i^2}{\sum M_i^2 / W_i^2}$$

        where $M_i = (|Z_{1i}| + |Z_{2i}|)/2$

                $\phi_i = $ phase $(Z_{1i} Z_{2i}^{*})$

                $^{*} = $ complex conjugate

                $W_i = $ coil gain factor

        18.   The combination as defined by claim 17 and
further including means for imaging velocity in a region using
said magnetic multicoil image, M.

1/2

H_1

10

B_0

10

14

12

FIG. 1A

Y

Z

X

FIG. 1B

Y

Z

X

FIG. 1C

Y

Z

X

FIG. 1D

24 — TRANSMITTER

22

GRADIENT AMPLIFIER

26

RF COIL

GRADIENT COILS

RF COIL

26

MAGNET

Y

Z

X

RECEIVER

28

DIGITIZER

30

OPERATING CONSOLE

KEYBOARD

CRT

FUNCTION BUTTONS

TRACK BALL

COMPUTER — 20

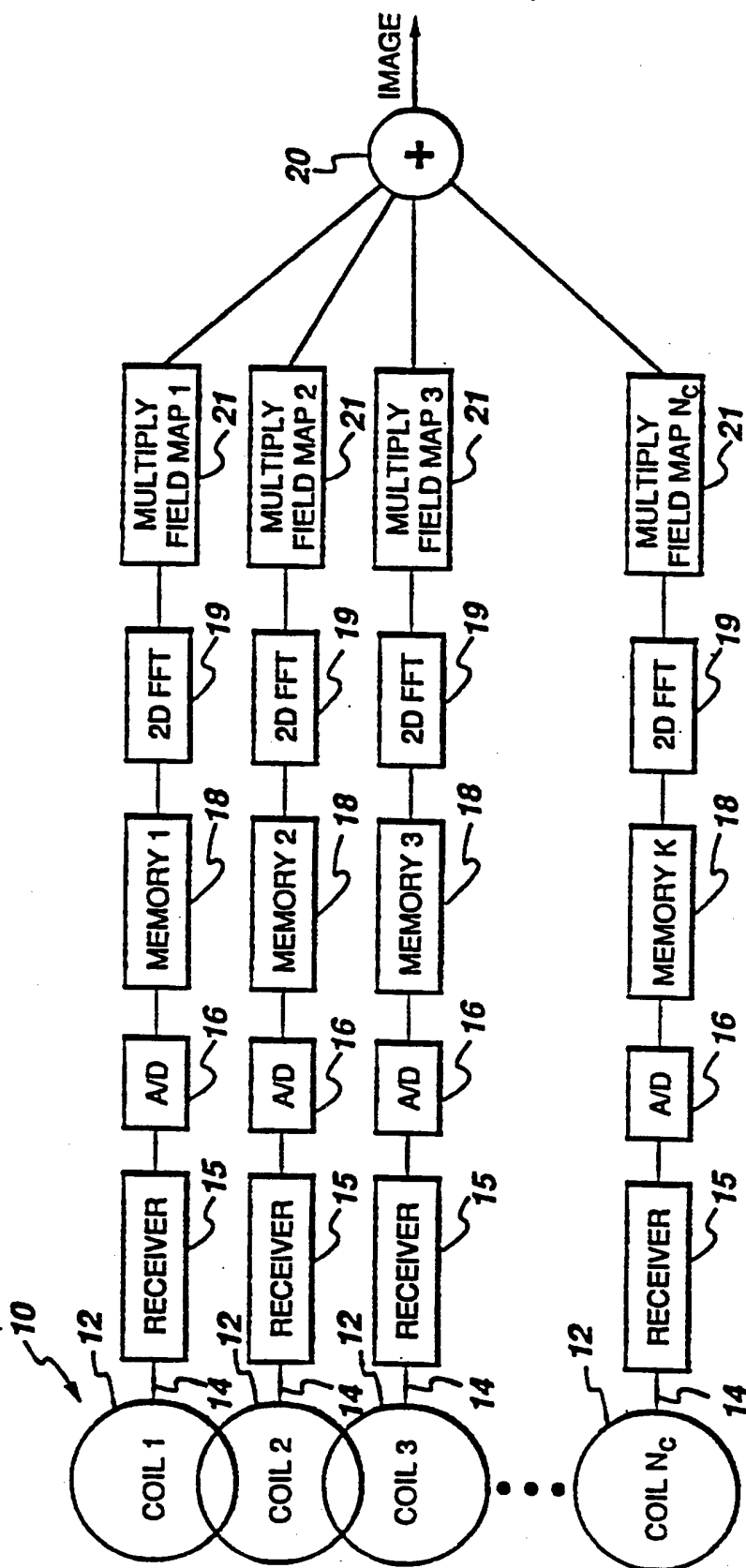MULTI-FORMAT CAMERA

MAGNETIC TAPE UNIT

FIG. 2

Fig. 3

INTERNATIONAL SEARCH REPORT

International application No.

PCT/US94/08954

A. CLASSIFICATION OF SUBJECT MATTER

IPC(5) :G01V 3/00
US CL :324/309

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 324/309, 307, 300, 318, 322, 306
128/653.2, 653.3

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US, A, 5,086,275 (ROEMER) 04 February 1992, see Fig. 1, summary of the invention and claims	1-18
A	US, A, 4,871,967 (ROTEM ET AL) 03 OCTOBER 1989, see abstract of the disclosure	1-18
A	US, A, 4,780,675 (DeMEESTER ET AL) 25 October 1988, see Fig. 1 and corresponding decription.	1-18

☐ Further documents are listed in the continuation of Box C.   ☐ See patent family annex.

*	Special categories of cited documents:	"T"	later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"A"	document defining the general state of the art which is not considered to be part of particular relevance		
"E"	earlier document published on or after the international filing date	"X"	document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"L"	document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)	"Y"	document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"O"	document referring to an oral disclosure, use, exhibition or other means		
"P"	document published prior to the international filing date but later than the priority date claimed	"&"	document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
15 SEPTEMBER 1994	DEC 06 1994

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